Gluteus minimus: an intramuscular EMG investigation of anterior and posterior segments during gait

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Abstract

Gluteus minimus is believed to consist of two structurally and functionally unique segments (anterior and posterior); however there is a lack of electromyography (EMG) research that attempts to verify current theoretical knowledge of this muscle. The purpose of this study was therefore to evaluate the function of gluteus minimus during gait, and to determine whether anterior and posterior segments are functionally independent. Bipolar fine wire intramuscular EMG electrodes were inserted into anterior and posterior gluteus minimus segments of fifteen healthy volunteers (9 males) according to previously verified guidelines. Participants completed a series of four walking trials, followed by maximum voluntary isometric contractions in five different positions. Temporal and amplitude variables for each segment were compared across the gait cycle with independent t-tests. The relative contribution of each segment to the maximum resisted trials were compared with Mann-Whitney U tests ($\alpha = 0.05$). Anterior and posterior segments were contracting at different relative intensities for three of the five maximum resisted trials (effect size = 0.39 to 0.62, $P < 0.037$). The posterior segment was larger in EMG amplitude (peak and average) during the first 20% of the gait cycle (effect size = 0.96 to 1.03, $P < 0.02$), while the anterior segment peaked later in the stance phase (effect size = 0.83, $P = 0.034$). Gluteus minimus is therefore composed of functionally independent segments. These results build on contemporary theoretical knowledge and may signify hip stabilising roles for each segment across different phases of the gait cycle.

Keywords: Hip, Buttocks, Electromyography, Gluteus minimus, Gait
1 **Introduction**

Gluteus minimus (GMin) is believed to have a pivotal role in hip joint stability [1-3]. The anterior component of this fan shaped muscle is directed vertically [1, 2, 4] with some fibres attaching onto the antero-superior capsule [2, 5] which may help to minimise suprolateral translation of the head of femur (HOF) during gait [2]. The posterior fibres are directed almost horizontally [1, 2, 4, 5] and are proposed to draw the HOF into the acetabulum and further facilitate this stabilising role at the hip joint [1-3]. There is some suggestion that these uniquely oriented segments have potential for independent function [1, 2], otherwise termed “muscles within muscles” [6].

Most of our understanding of GMin function has been inferred from cadaveric studies [1-3], biomechanical modelling [7], and magnetic resonance imaging (MRI) [8-10]. Functional assumptions from cadaveric research and biomechanical modelling are often provided in the context of gait; however there are no studies that have verified these assumptions *in-vivo* with electromyography (EMG).

The first and only EMG investigation of GMin occurred over thirty years ago [11]. One fine wire electrode was inserted “two inches” posterior to the anterior superior iliac spine (ASIS) in thirteen participants. A qualitative assessment of EMG amplitude was recorded from integrated EMG signals using a grading scale after participants performed a series of muscle contractions. The investigators concluded that GMin may act as an abductor, flexor, internal rotator or extensor of the thigh. While innovative at the time, contemporary EMG analysis and processing has gone well beyond qualitative descriptions of EMG signals. Furthermore, the use of a single electrode without verification of the location does not allow for the interpretation of the potentially independent segments of the muscle.
There is a clear lack of verified evidence to support current theoretical assumptions of GMin function, particularly in the context of gait.

Recent work has verified an EMG protocol and intramuscular insertion techniques to allow for the collection of muscle activity data from anterior and posterior GMIn segments [4, 12]. The aims of the current study were to generate a segmental EMG profile of GMIn during gait to evaluate theoretical models of GMIn function in the context of gait, and to determine whether GMIn is composed of functionally distinct segments (anterior and posterior).
2 Methods

2.1 Participants

Fifteen healthy, active adults (9 male and 6 female) aged 18 to 27 years were recruited for this study (Table 1). Participants were excluded if they had sustained a back or lower limb injury in the last six months or had a history of congenital hip disease or surgery on the hip or lumbar spine. To ensure an active cohort was recruited, participants were required to be involved in at least two hours of weight-bearing and sweat inducing exercise per week. Ethical approval was granted by the University Human Ethics Committee (Ethics ID: UHEC 10-065) and all participants provided informed consent.

Insert Table 1 here

2.2 Instrumentation and electrode insertions

Two bipolar fine-wire intramuscular electrodes were prepared [12] and inserted according to previously verified guidelines [4, 12]. Briefly, anterior and posterior GMin insertion sites were marked on the stance leg of participants [13] with reference to major surface landmarks [4]. Electrodes were inserted with the aid of real-time ultrasound (RTUS) imaging (HDI 3000; Advanced Technology Laboratories, Washington, USA) [12], and Doppler functionality to avoid the superior gluteal neurovascular bundle overlying posterior GMin [4].

A 5 cm Dermatrode reference electrode (American Imex, CA, USA) was positioned over the dorsum of the contra-lateral hand. Footswitches (Model: 402, Interlink Electronics, California, USA) were used to record temporal aspects of the gait cycle as described
previously [14]. A Delsys® Bagnoli-16 EMG system (Delsys Inc., Boston, USA) was used to record the raw signal from the footswitches and intramuscular electrodes.

2.3 Experimental protocol

The gait trials of this study were performed according to previous EMG gait research [15]. Before the walking trials, participants were given a 3-minute warm up to acclimatise to the testing protocol. Participants were then asked to walk barefoot, at a self-selected, comfortable walking speed along a 9 m walkway. This was repeated six times, of which the final four trials were recorded for analysis. Trials (timed with a stop-watch) were repeated if they exceeded ± 5% of their average walking speed (established during warm-up).

Following the gait trials, participants were asked to perform a series of maximum voluntary isometric contractions (MVICs). It has been recommended that multiple tests be performed in order to obtain the optimum maximum value for a muscle’s MVIC for subsequent amplitude normalisation [16]. Pilot testing revealed that open chain hip flexion, external rotation and abduction in external rotation were least likely to record a true maximum for any GMin segment [17] and were therefore excluded from testing to minimise participant fatigue. The MVIC positions tested were open chain hip abduction, hip internal rotation, hip abduction in internal rotation, hip extension and the clam exercise (moving knees apart while keeping feet together in a position of 45° hip and knee flexion). Each MVIC trial was performed three times for three seconds, with a three minute respite between each trial as described in detail previously [14].
2.4 **EMG data processing and analysis**

The raw EMG signal (Fig. 1A.) was passed through a differential amplifier at a gain of 1000 with a sampling frequency of 2kHz. A band pass filter (built into the amplifier: Delsys Inc., Boston, USA) of 20-2000Hz was applied. To reduce low frequency movement artefact, with minimal interruption to the raw EMG signal, the raw signals were passed through a 4th order high-pass Butterworth filter with a cut-off level of 50 Hz [14, 18]. Signals were then full wave rectified and further filtered with a low-pass 4th order Butterworth filter at a cut off level of 6 Hz with phase lag to generate a linear envelope [14, 19] (Fig. 1B.). This procedure was applied to gait and MVIC signals. Gait signals were then amplitude normalised to % MVIC, and time normalised to 100 points (% gait cycle).

Two consecutive strides representing the two middle strides of each walking trial were further processed for analysis (2 strides x 4 trials = 8 strides per participant) [15]. Data collected from the middle two strides ensured participants were not accelerating or decelerating at the point of analysis. For each muscle segment and participant, an ensemble average was generated from the eight strides. All participants’ ensemble averages were summed and averaged to produce a grand ensemble for GMin anterior and posterior, and establish an EMG profile for each segment across the gait cycle. Consistent bursts of EMG activity were identified in the grand ensemble curve at early stance (0%-20% gait cycle) and mid to late stance (20%-60% gait cycle). Data were therefore acquired from three phases of the gait cycle: 0% to 20%; 20% to 60% and total stance (heel strike to toe-off).
Delsys EMGworks 4.0 signal analysis software was used to acquire the dependant variables from each phase of the gait cycle. These were established from the linear envelopes of each participant’s trials. For each muscle segment, values were obtained for peak amplitude (% MVIC), average amplitude (% MVIC) and time to peak (TTP, % of gait cycle) from each phase of the gait cycle (0-20%, 20-60%, and total stance).

Data from the five MVIC positions were used for amplitude normalisation of gait variables, and for further comparisons between anterior and posterior segmental function. The mean EMG amplitude during an MVIC was calculated from the middle 1s of each MVIC trial. The highest amplitude value across all five positions was considered MVIC for each segment and for each participant.

The reliability of data processing was determined by re-processing data from 5 randomly chosen participants, 18 months apart. The intraclass correlation coefficients (ICC2,1 = 0.965-1.000) indicate excellent intra-rater reliability for temporal and amplitude EMG variables recorded across all phases of the gait cycle; the MVIC value used for amplitude normalisation; and onset detection from footswitch signals.

The temporal and amplitude gait variables from each segment in each phase (0% to 20%, 20% to 60%, and total stance) were used for quantitative comparisons. Histograms and the Kolmogorov-Smirnov (K-S) test were used to explore the assumption of normality within these variables and where normality could not be assumed, variables were log-transformed and re-assessed for normality [20]. Independent samples t-tests compared the means of anterior and posterior segments across all gait variables. To provide an estimate of the magnitude of difference (effect size) between segments, a standardised mean difference
SMD = mean difference / pooled SD) was calculated for all gait comparisons [21]. An effect size threshold of 0.2, 0.5 and 0.8 was considered small, medium and large respectively [22]. Segmental comparisons for MVIC variables were performed with the Mann-Whitney U test. A standardised effect size for this test was calculated by dividing the z-score of the Mann-Whitney U test by the square root of the total sample size [20]. All statistical analysis were performed in SPSS (version 19, IBM SPSS Inc., Chicago, IL, USA) using an alpha of 0.05.
3 Results

Participant details can be seen in Table 1. Data from one participant’s posterior GMin electrode was not processed as it was dislodged during the testing process. Data from 14 posterior GMin segments and 15 anterior GMin segments remained. The mean (SD) walking speed and stride time was 1.17 (0.15) m s$^{-1}$ and 1.04 (0.11) s respectively.

Insert Table 1 here

3.1 Gait

The grand ensemble curves for anterior and posterior GMin illustrate a biphasic activation pattern during the stance phase of gait (Fig. 2.). The first burst occurred within the first 20% of the gait cycle and the second burst within the 20% to 60% phase. The grand ensemble also suggested that the amplitude of the second peak of anterior GMin was on average greater than its first peak, and this was the case in 10 out of 15 participants. In contrast, the amplitude of the second peak of posterior GMin was on average lower than its first peak, and this was the case in 9 out of 14 participants.

Fig. 2 also illustrates qualitative comparisons between anterior and posterior GMin segments, while quantitative comparisons are presented in Table 2. Posterior GMin had a significantly higher peak ($P=0.02$) and average amplitude ($P=0.01$) than anterior GMin in the first 20% of the gait cycle. There were no significant differences between segments for any variable within the 20% to 60% phase ($P>0.05$). When the total stance phase was considered, anterior GMin had a significantly lower peak amplitude ($P<0.05$), and later TTP ($P=0.04$) than posterior GMin. All significant gait findings were large in magnitude (ES > 0.80).
3.2  **MVIC**

Segments were contracting at significantly different intensities during internal rotation (anterior > posterior; small to moderate ES), abduction in internal rotation (posterior > anterior; small to moderate ES), and clam (posterior > anterior, moderate to large ES) (Table 3).
4 Discussion

This is the first study to illustrate the EMG profile of GMin within the gait cycle (Fig. 2). It is also the first study to use verified EMG guidelines for assessing segmental function of GMin. The results suggest that GMin posterior has its greatest activity early in the gait cycle (0% to 20%), while GMin anterior consistently peaks later in the gait cycle (20% to 60% phase) (Fig. 2 and Table 2). The gait and MVIC data also suggest that anterior and posterior segments can function independently (Table 2 and 3).

The one prior EMG investigation of GMin [11] investigated muscle activity during resisted and un-resisted hip motion in a recumbent position. Based on a qualitative analysis, the authors concluded that GMin is an internal rotator and abductor of the thigh; it can be an extensor or flexor of the thigh depending on which fibres were activating; and does not externally rotate the thigh. The MVIC data from the current study is consistent with this, demonstrating GMin can be considered an internal rotator and an abductor of the thigh. In addition to the previous findings, the GMin was also active at very high levels during maximum resisted abduction in internal rotation. In contrast to prior conclusions the current findings suggest that GMin is highly active during maximum resisted thigh extension in the neutral hip position, regardless of which fibers are being assessed.

4.1 Muscles within muscles

In combination, the gait and MVIC data indicate that GMin is comprised of two functionally distinct segments. For example, during the clam MVIC manoeuvre, anterior GMin was active at low levels (mean 10.8% MVIC), while posterior GMin was active at moderate levels (mean 48.2% MVIC). The difference between the two segments was moderate to large, and statistically significant. This is the first study to conclusively report
that GMin is composed of functionally unique segments. Future research of GMin in healthy or clinical populations must therefore consider recording data from each segment so as not to generalise information from one independent segment to the whole muscle.

4.2 GMin function during gait

The functional role of GMin and its segments has been inferred from cadaveric specimens, however these finding have not previously been validated with dynamic gait studies. It is generally agreed that GMin has a fundamental role in hip joint stability [1-3], as the arrangement of GMin’s fascicles parallel to the neck of femur (NOF) are aligned to draw the head of femur (HOF) in a superior-medial direction towards the acetabulum. This is believed to contribute to compressive hip joint contact forces, facilitating its femoral head stabilising role. Biomechanical modelling indicates that muscles contribute to 95% of the superior and medial contact forces across the hip joint during the gait cycle, with GMin and gluteus medius (GMed) being the major contributors [7]. Both of these contact forces have two peaks, one at contra-lateral toe-off (≈ 18% gait cycle) and the other just prior to contra-lateral heel strike (≈ 45% gait cycle). Given the current findings that GMin posterior is significantly more active than anterior GMin during the first burst in early stance, we propose that it is a major contributor to superior-medial contact forces in this phase of the gait cycle, supporting its role as a primary femoral head stabiliser in early stance [1]. The second peak in superior and medial contact forces [7] corresponds with a reduction in EMG activity of GMin posterior and an increase in EMG activity of GMin anterior, resulting in a relative co-contraction of both segments during late mid-stance (second burst, Fig. 2). Therefore, the primary femoral head stabilising role may be attributed to posterior GMin in early stance [1], and a co-contraction of posterior and anterior GMin in late mid-stance [2].
The role of anterior GMin in late mid-stance may be two-fold. First, biomechanical modelling suggests that anterior hip joint forces increase with greater hip joint extension [23]. The fascicle arrangement of anterior GMin, together with its close attachments to the hip joint capsule [2, 5] places it in a unique position to stabilise the HOF in the acetabulum, and together with iliopsoas, minimize anterior hip joint forces during mid to late stance [23]. Anterior GMin may therefore have a role in reducing the potential stresses placed on hip joint ligaments, the anterior hip joint capsule and the anterior superior acetabular labrum [23]. Second, the MVIC data indicate that anterior GMin is very highly active during hip internal rotation (86% MVIC, Table. 3). This is also supported by cadaveric studies and biomechanical models suggesting that anterior GMin has a large internal rotation torque producing potential [24]. Therefore, with the lower limb fixed, GMin anterior can potentially contribute to forward rotation of the contra-lateral pelvis during the stance phase of gait. However, other reports suggest that this role may be better attributed to anterior GMed [1, 3, 14], given its larger physiological cross-sectional area [5], and more favourable internal rotation moment arm [25]. Further EMG work with kinematic and kinetic data will help to clarify these speculations.

4.3 Clinical implications

Knowledge of segmental muscle activity within GMin, as well as where the relative peaks of each segment occur across the gait cycle may have a number of clinical implications. For example, analysis of TTP across stance demonstrated that anterior GMin peaked more consistently during the second burst. This suggests that anterior GMin has a greater contribution as the weight bearing hip extends, later in the gait cycle when compared with posterior GMin. This is potentially an important clinical finding, as people who suffer
conditions that result in reductions in stride length (e.g. following THA [26]) may develop specific and localised atrophy of anterior GMin (as identified in THA [9]). This knowledge could be used to help develop specific and targeted rehabilitation programs for strengthening this muscle segment in functionally meaningful positions i.e. in positions of greater hip joint extension.

4.4 Limitations

The limitations of this study primarily relate to the generic limitations of intramuscular EMG and MVIC normalisation. Fine wire electrodes record activity from a small sample of muscle fibres and it is assumed that this is representative of the entire segment. There is some debate about the best amplitude normalisation procedure that should be considered, with MVIC normalisation resulting and large between subject variability [18], and may reflect the large SD’s presented in Table 2 and 3. However MVIC normalisation has commonly been preferred as it provides information about the intensity of a muscle contraction relative to its maximum capacity, offering a clinically relevant amplitude scale [27], but this assumption has limitations [28]. Furthermore, test-retest reliability has not been established for this protocol.

5 Conclusion

This study has addressed the lack of EMG research into GMin, and is the first study to generate an EMG profile of GMin during gait. We conclude that GMin is indeed composed of “muscles within muscles” whereby unique functional properties have been identified within uniquely oriented anterior and posterior segments. Future work on GMin must consider the function and clinical relevance of each structurally unique segment.
6 Conflict of interest

None declared


Fig. 1. Schematic illustration of EMG signal processing for one participant across one full stride. A: Raw un-processed anterior gluteus minimus EMG signal. B: Processed EMG signal representing the EMG profile of anterior GMin across the gait cycle. Dotted horizontal white arrow represents peak amplitude of the second burst; solid vertical white arrow represents time to peak amplitude of the second burst. MVIC, maximum voluntary isometric contraction; TO, toe-off.

Fig. 2. Grand ensemble EMG averages (solid lines) with 95% confidence intervals (CI’s, dotted lines) for anterior (15 participants) and posterior (14 participants) gluteus minimus (GMin) across the gait cycle. Differences between anterior and posterior GMin muscle activity are likely to occur where 95% CI’s do not overlap. Vertical dashed line indicates mean toe-off (62%). Note, peak bursts in this figure represent mean peak activity within and across participants, therefore do not reflect absolute peak values of each burst in Table 2.